(19) (CA) **CANADIAN PATENT** (12)

(54) X-Ray Imaging System and Solid State Detector
Therefor

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TITLE OF THE INVENTION

X-RAY IMAGING SYSTEM AND
SOLID STATE DETECTOR THEREFOR

BACKGROUND OF THE INVENTION

5   Field of the Invention

This invention relates to x-ray imaging systems
and particularly to a x-ray imaging system which
utilizes a solid state x-ray detector.

Discussion of Related Art

10   Presently, x-ray imaging systems are utilized
in a variety of applications, both as medical
diagnostic tools and for industrial quality control.
The most common form of x-ray detection resides in the
use of silver halide film. However, the use of such
15   film requires the performance of several wet, control
requiring chemical developing steps. In addition, this
film is expensive, thus increasing the cost of x-ray
images produced in this manner.
It would be highly desirable, therefore, to
20   produce an x-ray imaging system which does not require
the use of silver halide film. Several detectors have
been proposed for this purpose.
For example, U.S. Patent No. 4,471,378 to Ng
discloses a light and particle image intensifier which
25   includes a scintillator and photocathode unit for

- 1 -
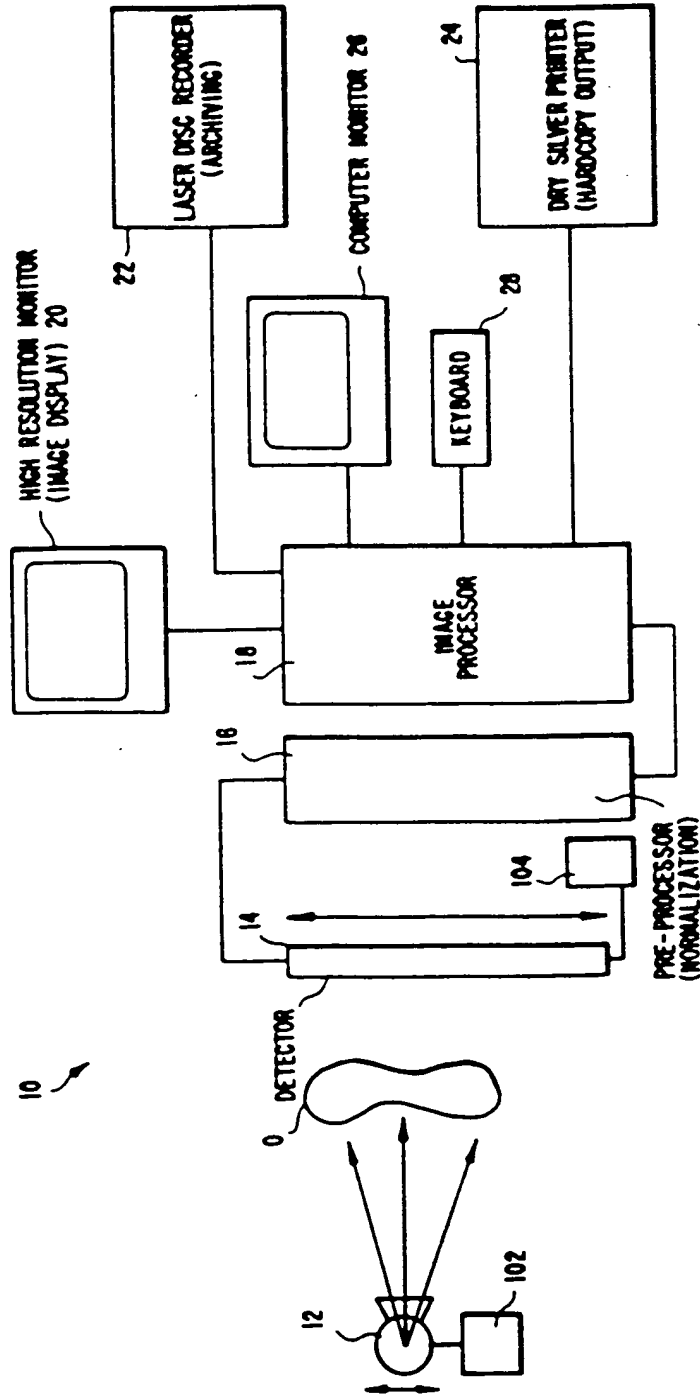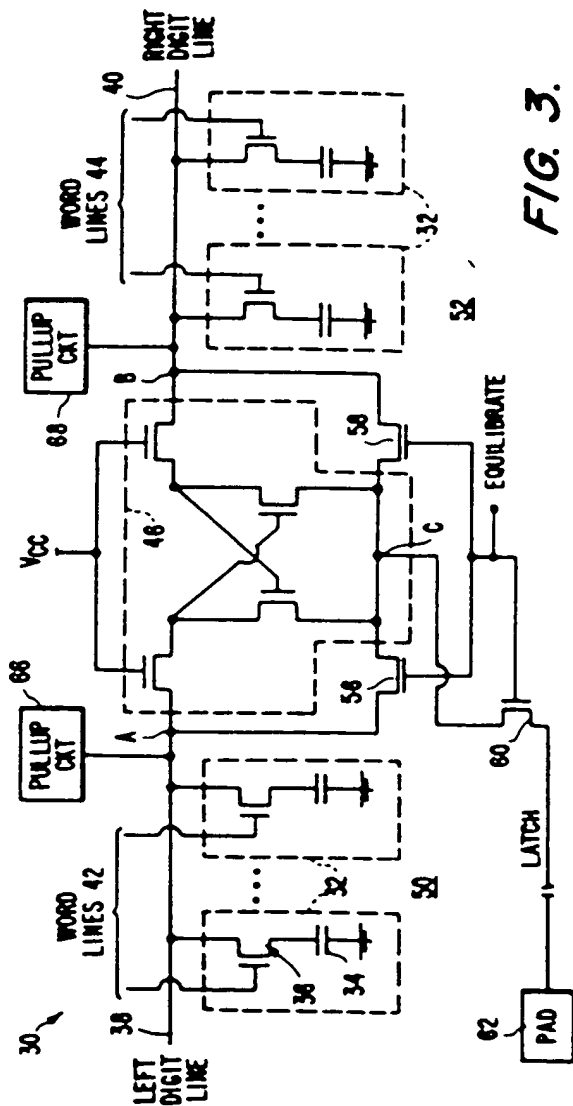
FIG. 1.

HIGH RESOLUTION MONITOR
(IMAGE DISPLAY) 20

LASER DISC RECORDER
(ARCHIVING)

22

COMPUTER MONITOR 26

IMAGE PROCESSOR

18

KEYBOARD

28

DRY SILVER PRINTER
(HARDCOPY OUTPUT)

24

16

14

DETECTOR

0

PRE-PROCESSOR
(NORMALIZATION)

104

12

102

10

FIG. 2.

LEFT DIGIT LINE

WORD LINES 42

PULLUP CKT 68

Vcc

PULLUP CKT 68

WORD LINES 44

RIGHT DIGIT LINE

EQUILIBRATE

LATCH

PAD

FIG. 3.
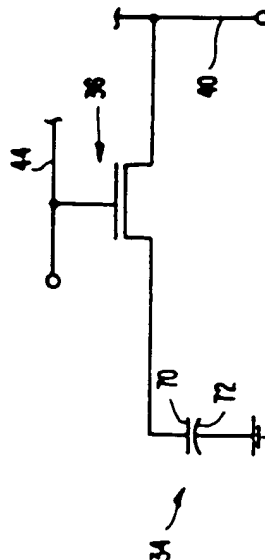
FIG. 4.
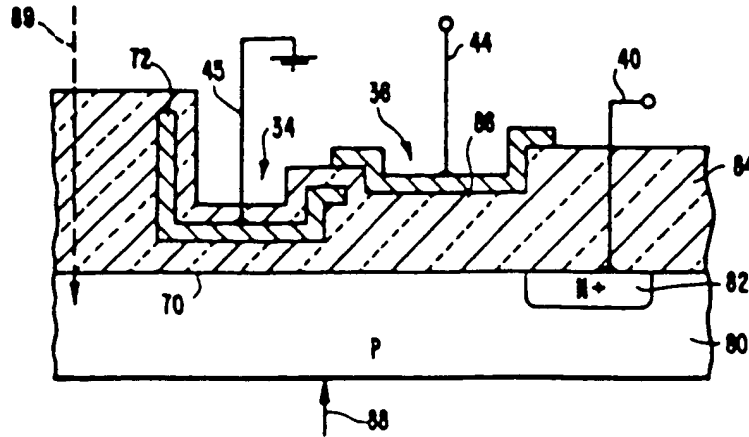
FIG. 5.
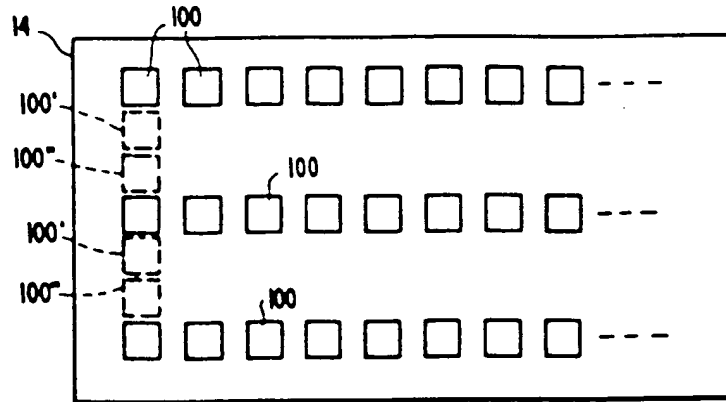
FIG. 6.
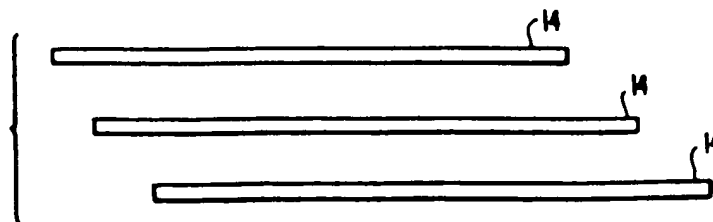
FIG. 7.

FIG. 9.

PATENT AGENTS

Swabey, Mitchell, Houle,
Marcoux & Sher.

FIG. 8.

```
        ( START )────120

          INITIALIZE      ────122
          SET m·1 ! n·1

          READ  f(m,n)    ────124

          LOOK UP f(m,n)  ────126

          DETERMINE
          f(m,n) GRAY SCALE  ────128
          INTERVAL  j

          LOOK UP f*(j)   ────130

          f*(m,n) · f*(j) f(m,n)  ────132

          DETERMINE GRAY
          SCALE INTERVAL   ────134
          FOR f*(m,n)

          ASSIGN GRAY
          SCALE TO (m,n)   ────136

  140
  INCREMENT                ALL
  m,n          NO────<    CELLS  >────138
                          READ
                           │ YES
  142                     ( END )
```

PATENT AGENTS

Awabery, Mitchell, Houk,
Macoux & Aker.
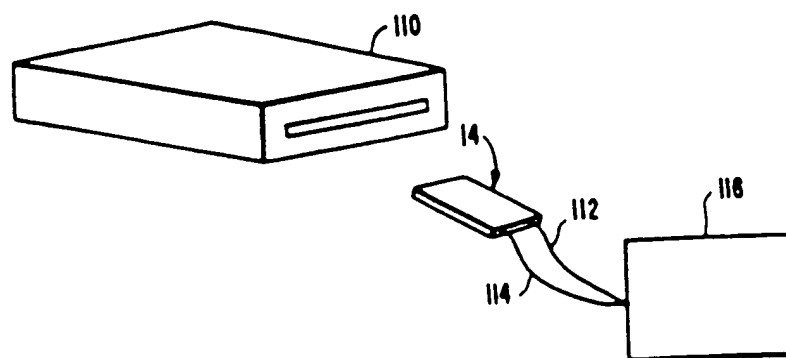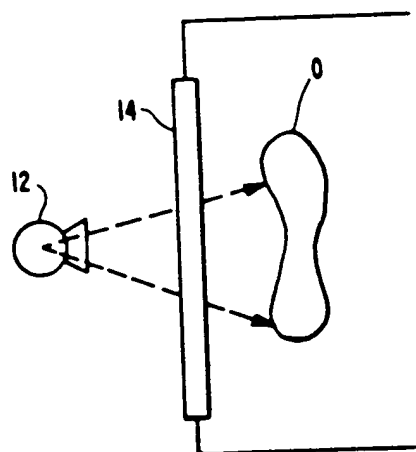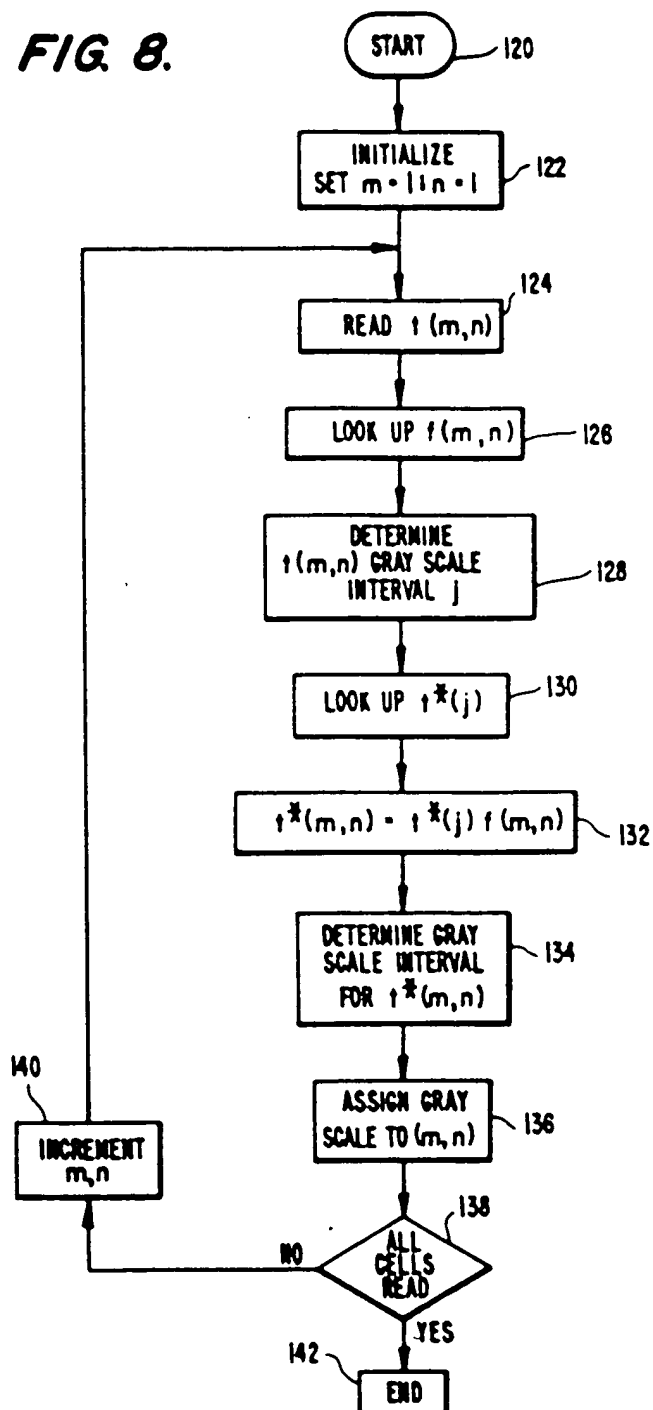
converting incident image conveying light or charged
particles to photoelectrons and a charge coupled device
for detecting the photoelectrons and transmitting to
data processing and video equipment information

5   relating to the quantity or energy level as well as the
location of the electrons impinging on the sensing
areas of the charge couple device.

U.S. Patent No. 4,413,280 to Adlerstein et al
discloses an x-ray imaging apparatus which includes a

10 transducer for converting incident x-radiation to a
corresponding pattern of electrical charges. The
charges generated by the transducer are accelerated
onto an array of charge detecting or charge storing
devices which store the charges in the form of an

15 electrical signal corresponding to the charge pattern.

SUMMARY OF THE INVENTION

One object of the present invention is to

20 provide a solid state imaging system and detector which
are highly sensitive to x-radiation and can produce
highly accurate x-ray images.

Another object of the present invention is to
provide an x-ray imaging system and detector which can

25 be produced by conventional solid state fabrication
technology.

A further object of the present invention is to
provide a solid state imaging detector which can be
produced in such small sizes as to enable its use in

30 very confined areas.

Another object of the present invention is to
provide an x-ray imaging detector which can be
substituted directly for x-ray film used in
conventional x-ray imaging systems.

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Yet another object of the present invention is provide an x-ray detector which is relatively inexpensive to fabricate so as to enable its use in fixed locations for ease of periodic x-ray analysis of mechanical structures and the like.

In accordance with the above and other objects, the present invention is an x-ray imaging system comprising an x-ray source for producing an x-ray field, and an x-ray detector. The x-ray detector comprises a solid state integrated circuit having a plurality of charge storage devices and a circuit for placing a charge on the charge storage devices. The charge storage devices are disposed in an x-ray permeable material and the detector is positioned in the x-ray field such that the charge is dissipated by secondary radiation produced by interaction of the x-ray field in the silicon substrate of the solid state integrated circuit.

The charge storage devices may be divided into groups to form pixels. Each pixel comprises one or a plurality of charge storage devices and the exposure times for discharging the charge storage devices in a single pixel can be different from one another to provide a gray scale.

In accordance with other aspects of the invention, the integrated circuit may be a dynamic random access memory.

Each charge storage device comprises a single cell of the integrated circuit. The cells are spaced from each other such that dead space exists therebetween. Also, the cells are produced in banks of 32,000 with about 1/4 mm dead space between banks. A plurality of detectors may be stacked with the cells of the detectors staggered such that each cell of one detector is positioned behind the gap between cells of another detector so as to eliminate all dead space.

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The imaging system also includes processing
circuitry for accessing the cells of a detector. The
processing circuitry may include a system for
normalizing the outputs of all of the cells to
5   compensate for various inherent differences in
radiation sensitivities of the various cells.

BRIEF DESCRIPTION OF THE DRAWINGS

The above and other objects of the present
invention will become more readily apparent as the
10   invention is more clearly understood from the detailed
description to follow, reference being had to the
accompanying drawings in which like reference numerals
represent like parts throughout, and in which:

Figure 1 is a block diagram of the x-ray
15   imaging system of the present invention;
Figure 2 is a circuit diagram of an integrated
circuit detector used in the imaging system of Figure
1;
Figure 3 is an enlarged schematic showing one
20   charge storage capacitor of the circuit diagram of
Figure 2;
Figure 4 is a cross section of a chip showing
the structure depicted schematically in Figure 3;
Figure 5 is a view of a portion of the detector
25   of the present invention stacked over additional
detectors to fill up the dead space between cells;
Figure 6 is an end elevational view of the
stacked detectors of Figure 5;
Figure 7 is a diagrammatic representation
30   showing the system of the present invention used in
place of x-ray film; and

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Figure 8 is a flow diagram depicting a method
of normalizing the cells of the present invention.

DESCRIPTION OF THE PREFERRED EMBODIMENT

Figure 1 shows the x-ray system 10 to comprise
5   an x-ray source 12 and a detector 14 positioned to
receive the radiation from source 12. Source 12 can be
any standard x-radiation source such as manufactured by
G.E. or Siemens, or can be an ultra small focal spot
source such as manufactured by Ridge or Magnaflux. Any
10   size focal spot source can be used. Currently, the
smallest focal spot available is one micron. Also, the
source 12 and detector can be placed as close to the
object O to be x-rayed as desired due to the
configuration of the detector 14, as will become
15   readily apparent.
Detector 14 may be a dynamic random access
memory such as the IS32 OpticRAM sold by Micron
Technology, Inc. of Boise, Idaho. This device is an
integrated circuit DRAM having 65,536 elements and is
20   used as a solid state light sensitive detector. The
Micron DRAM is specifically adapted to sense light
inasmuch as there is no opaque surface covering the
integrated circuit. However, any type of dynamic
random access memory may be used for detector 14 as
25   long as the covering is transparent to x-radiation. In
fact, as will become apparent, any type of dynamic
memory element may be used as detector 14. The memory
element does not have to be a random access memory,
although the use of a random access memory facilitates
30   preprocessing and image processing routines.
The output of detector 14 is passed to
preprocessor 16 which serves the function of·
normalizing the outputs of all of the cells of detector
14. That is, the sensitivity of the cells of detector

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14 will inherently vary. A normalization value can be
stored in preprocessor 16 so as to normalize the output
of each of the cells to ensure a field describing
reading.

5   The normalized output of preprocessor 16 is
passed to image processor 18 which manipulates the data
using conventional image processing programs as well as
new image processing programs which will be made
possible by the present invention, such as "zoom"

10 programs which are not currently in existence. This
image can be displayed on a high resolution monitor 20,
can be stored on a laser disc recorder 22, can be
printed using a dry silver printer 24, or can be sent
via satellite to remote image processors (not shown).

15 A menu driven program is displayed on a computer
monitor 26 prompting appropriate instructions and data
which can be entered into the image processor 18 using
a keyboard 28.

       Figure 2 shows a schematic diagram of a portion

20 of a typical DRAM used in detector 14. The circuit 30
comprises a plurality of cells 32, each of which
contains a memory capacitor 34 and an access transistor
36. The individual cells are accessed through left and
right digit lines 38 and 40, respectively, as well as

25 word lines 42 and 44. A sense amplifier 46 is provided
in the form of a cross coupled MOSFET detector circuit.
The sense amplifier 46 has nodes A and B which are
coupled to the left digit line 38 and to the right
digit line 40, respectively. The cells 32 are divided

30 into a left array 50 and a right array 52. The left
array 50 is accessed by the left digit line 38 and the
right array 52 is accessed by the right digit line 40.
The word lines 42 access the individual cells of array
50 and the word lines 44 access cells of the array 52.

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A pair of equilibrate transistors 56 and 58
couple the digit lines together to allow equalization
of the digit lines at the end of a refresh cycle and
during the precharge state of the next cycle.

5 The common drains of the cross coupled sense
amplifier transistors at node C are connected through
an isolation transistor 60 to a pad 62 on the periphery
of the integrated circuit chip. The pad 62 is bonded
to one of the leads of the circuit chip package.

10 A pair of pull up circuits 66, 68 are coupled,
respectively, to the nodes A and B. The pull up
circuits 66, 68 are voltage divider circuits operable
to control the voltage level of the digit lines 38 and
40.

15 Figure 3 shows one cell of the circuit 30. For
convenience, the cell is shown to be one of the array
52 but it could be any of the cells. As shown, the
capacitor 34 has two plates 70 and 72 between which a
charge is stored. Initially, the capacitor is charged

20 by applying a high potential on word line 44 and a high
potential on right digit line 40. This corresponds to
a "1" state of the cell. In the presence of incident
x-radiation, the charge on capacitor 34 is dissipated
as will be discussed below.

25 With reference to Figure 4, the portion of the
integrated circuit containing the cell shown in Figure
3 is set forth in cross section. The circuit comprises
a p-type silicon substrate 80 onto which an n+ region
82 has been added. A silicon dioxide layer 84 is

30 deposited over the substrate and n+ region 82 to form
an insulating layer. The lead 40 is connected to the
n+ region to form the drain of transistor 34. A metal
plate 86 is formed on the oxide layer 84 to form an
insulated gate of transistor 36. The capacitor 34 is

35 formed by a metal plate 72 and the interface 70 between
p-type substrate 80 and oxide layer 84, which forms the

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other capacitor plate.

When the cell of Figures 3 and 4 is set to the
"1" state, charge is built up on the interface 70 to
charge the capacitor 34. The gate voltage is then
lowered so as to discontinue communication between the
drain voltage at line 40 and the capacitor 34. This
charge is dissipated due to the absorbtion of x-ray
photons in the substrate 80. In Figure 4, the
direction of incident x-radiation is shown by the arrow
88. The x-radiation produces free electrons in the
substrate 80 either by photoelectric effect, Compton
scattering, or pair production. The number of
electrons produced through photoelectric effect is
negligible. Compton scattering and pair production
have the highest probabilities of producing free
electrons.

It is noted that interconnections between
components of a cell and between cells are provided on
the oxide layer of the semi-conductor. This is
indicated in Figure 4 by showing the leads 40, 44 and
45 extending out of the oxide layer. Such leads
represent the interconnections produced by the
metalization layer of an integrated circuit.

Irradiation of the cell 30 from either side
results in virtually all of the x-radiation being
received in the substrate 80 so that the total electron
production for any given energy level of radiation is
achieved. The free electrons produced by interaction
between the substrate and the x-radiation decrease the
charge at junction 70 and thus decrease the charge on
capacitor 34.

Compton scattering and the photoelectric effect
are the relevant interactions causing the appearance of
free electrons. The relative occurrences of the
different reactions depends on the energy of the x-ray.
There is little direction sensitivity except in the

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case of Raleigh scattering, which is predominantly
forward but also yields the least free electrons.

Referring again to Figure 2, it will be
understood that the circuit 30 is of the type which
5   employs a dynamic/active/restore sense amplifier of the
type described in U.S. Patent No. 4,397,002, issued
August 2, 1983 to Wilson et al, and U.S. Patent No.
4,291,392, issued September 22, 1981 to Proebsting.

In operation of circuit 30, during one cycle, a
10   given word line 42, 44 in Figure 2 is brought to a
logic one level to enable the addressed access
transistor 36. The respective cell capacitor 34 is
discharged into the appropriate digit line (e.g. digit
line 40 for a capacitor of array 32) changing its value
15   above the equalized value. Then, a latch signal from
the pad 62 becomes a logic low state to enable
operation of the cross coupled transistors and the
sense amplifier 46 during absence of the equilibrate
signal. The sense amplifier 46 responds to the latch
20   signal by reducing the opposite digit line (in this
case digit line 38) to a ground potential. The digit
lines are connected by input/output circuitry (not
shown), which provides a digital signal representing
the content of the selected memory capacitor 36. The
25   pull up circuits cause the right digit line to be
pulled up to the level of the supply voltage. At
approximately this time, the storage capacitor 36 which
has been connected to the bit line has been restored to
its original logic one state. The word line is then
30   returned to ground to isolate the charge on the
respective memory cell. The digit lines are then
permitted to go low and the equilabration signal
becomes a logic high to render the equilibrate
transistors 56, 58 conductive to allow the digit lines
35   to be connected for equilization. This permits the
charge on the digit lines 38, 40 to be shared such that

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the digit lines equilibrate to a voltage approximately
half way between the supply voltage and ground. A new
cycle is thereupon ready to commence.

The latch signal is placed on pad 62 such that
during the equilibrate signal, a voltage potential may
be applied to the digit lines when they are connected
to effect equilibrium. As will become apparent, the
voltage which is applied to the pad 62 allows
adjustment to the sensitivity of the image sensor. In
particular, the digit line potential acts as a
threshold to determine if a particular memory cell 36
is at a high or low voltage level. By raising the
potential, cells may leak less before they are
considered to be decayed from a logic one to a logic
zero value. Accordingly, the sensitivity of the cells
can be adjusted by adjusting this potential.

Another way of adjusting the sensitivity of the
cells is by etching in different thicknesses of oxide
material in the capacitor 34 shown in Figure 4. In
other words, the thickness of the oxide between
interface 70 and metal plate 72 determines the
sensitivity of the capacitor to x-radiation. The
thicker this layer, the more sensitive the capacitor is
to radiation discharge. Accordingly, to make a more
sensitive detector, the layer should be made thicker
and to make a less sensitive detector, the layer should
be made thinner. There are, however, limitations to
this technique because the impedance of the capacitor
must lie in a certain range in order for the circuit to
function properly (varying the thickness changes the
impedance).

A further technique for adjusting sensitivity
is to adjust the exposure time of the detector cells.
Using conventional x-ray source controlling equipment,
the response of the cells can be adjusted either by
using higher energy x-rays or by increasing the

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intensity of the x-rays. Clearly, in either case, the cells will react more quickly.

The cells of the detector are binary in nature. In many applications, it is useful to have a gray scale. This can be accomplished using any of the three techniques discussed above for varying the sensitivity of the cells.

If the digit line potential threshold is varied to vary the sensitivity, a plurality of cells may be grouped to form a pixel. For example, 8 cells per pixel could provide a gray scale having 7 levels, although in practice, 80 cells per pixel (79 gray scales) may be used to provide redundancy at each level of the gray scale. In Figure 2, if it is assumed that array 32 has 8 cells, ideally, this array could act as a single pixel. During the cyclic operation of the circuit 30, as discussed above, a different digit line potential is applied to digit line 40 when reading each of the memory capacitors 34 of array 32. In this manner, the soak time required for each of the cells 32 to discharge to a low level would be different, thereby providing a gray scale.

The preprocessing circuit 16 of Figure 1 is programmed to provide the necessary variations in threshold potential, as would be apparent to one of ordinary skill in the art.

Likewise, if each of the cells of array 32 forms a single pixel, each memory capacitor 34 could be produced with a different thickness of oxide layer. In this manner, the capacitors would have inherent differing sensitivities, thus producing a gray scale. Clearly, the number of cells in a pixel can be varied to produce a gray scale having as many levels as desired.

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A third means of providing gray scale is to take several x-ray pictures with different soak times and to add the results. For example, with an object having a varying thickness, a plurality of images can be obtained using differing soak times. The lower soak times are used to depict the thinner portions of the object. By adding all of the images together, a multilevel gray scale image is obtained depicting the various thicknesses of the object.

The detector 14 can be produced in any desirable size. Conventionally, the largest silicon chips produced are 6" diameter circular wafers. The wafers are cut into individual integrated circuits and packaged as, for example, dual in-line packs. The detector 14 can be processed by conventional integrated circuit processing techniques and can be packaged in any conventional configuration such as the dual in-line pack.

Figure 5 shows a portion of a detector 14 having cells 100. The cells are each approximately 8 microns square and the cells of any row are spaced 9 microns from center to center. Thus, these cells are separated by only one micron. Vertically, as shown in Figure 5, the cells are spaced 25 microns from center to center. Also, the cells are arranged in banks of 32,000 with approximately 1/4 mm dead space between the banks. The dead space between cell banks is required to accommodate the trunk lines to the cells. The arrangement of cells in one integrated circuit is sufficiently dense to provide a very high resolution x-ray image for most applications. However, if an even higher resolution is required, since each chip is relatively transparent to radiation, a plurality of chips can be stacked and offset relative to one another to fill the gaps between cells and/or between the banks of cells. For example, as shown in Figures 5 and

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6, three detectors 14 are stacked and offset so that
the cells of the lower two detectors are positioned
between the cells of the upper detector. In Figure 5,
the cells of the lower detectors are indicated in

5   phantom as cells 100' and 100''. Accordingly, it can
be seen that a single detector can be built using three
integrated circuits in which virtually all of the
available area is filled with cells.

           As an alternative to stacking detectors,

10   several sequential images can be taken with either the
source 12 (Figure 1) or the detector 14 being moved
between each x-ray exposure. This motion is produced
by oscillating supports 102 or 104 shown in Figure 1.

           The integrated circuit itself is about 1/2 mm

15   thick. The entire detector can be made about 1/2 inch
thick, including all necessary connections etc.
Accordingly, a detector 14 can be substituted directly
for x-ray film in almost all existing x-ray devices.
As shown in Figure 7, the detector 14 can fit directly

20   into an x-ray film receptacle 110 with leads 112 and
114 being connected to the processing section 116 which
contains preprocessor 16, image processor 18, etc.
Processing section 116 may be a computer with the
preprocessor and image processor being software

25   components.

           Another advantage of the fact that the detector
14 can be produced in small sizes is that relatively
small detectors can be permanently located in virtually
inaccessible areas of structures to be x-rayed. For

30   example, structural elements in aircraft are required
to be examined periodically using x-ray techniques.
This normally requires disassembly of major structural
components. However, with the present invention, an
x-ray detector can be permanently mounted in place and

35   the detector leads can be made accessible through a
plug or the like. In this manner, x-ray analysis of

structural components can be carried out quickly and easily. Also, since the detector 14 is made using conventional integrated circuit techniques, its cost is relatively low thus also contributing to the feasability of utilizing a plurality of small x-ray detectors 14 on a component requiring x-ray analysis.

Furthermore, the small size of the x-ray detector 14 enables it to be placed in body cavities such as the ear or the like to facilitate clear clinical diagnostic x-ray images.

The fact that each cell of detector 14 is small in size enables the use of an x-ray source 12 having the smallest focal spot available. Also, since the cells are relatively closely packed, the present invention can be used in zoom radiography without requiring any change in the distance between the source and the detector. Normally, the size of an x-ray image can be increased by increasing the distance between the object and the detector. However, in the present invention, due to the large number of cells in the detector, the image can be electronically enlarged without significant loss of resolution by merely displaying the image on a larger fraction of the screen. Further, as discussed above, if additional resolution is required, detectors 14 can be stacked as shown in Figures 5 and 6.

Due to the small size of the cells of the invention, greater resolution can be achieved with little or no cross talk between the cells. That is, the length of travel of the most energetic secondary radiation produced by interactions between the x-ray beam and the silicon substrate 80 (Figure 4) is approximately 10 microns. Since this is also the separation distance between cells, there is very little probability that radiation impinging upon one cell will result in noise being produced in an adjacent cell.

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Furthermore, the present invention is suited for microradiography since the cells are small and closely packed. The resolution attainable using the invention is sufficiently high to permit an accurate representation of objects having a area on the order of 1,000 square microns.

Also, because of the digital nature of the present invention and its inherent compactness, it is possible to configure an x-ray imaging system in a unique way that permits the user to view the internal structure of objects where it is not possible or practical to place the imaging media behind the object. As shown in Figure 9, the x-ray source 12 is placed directly behind the detector 14 which can be placed directly on or in front of the object O to be viewed. In this configuration, the x-ray field produced by the source penetrates through the detector and impinges on the object. The image of the object is created in the detector by back scattered radiation from the object. The image produced in the detector 14 is that of the object O produced by back scattering superimposed on an image of the source 12 produced by the x-ray field passing through the detector 12 initially. The image of the source 12 can be removed by subtracting it from the total image. That is, an image of the source alone is produced by irradiation of the detector 14 with no object present and digitally subtracting this image from that of the combined object and source image.

It should be noted that it is preferable to orient detector 14 in Figure 9 such that the connection leads extend out of the side facing the source. In this way, the detector can be positioned closer to the object O and the image of the leads, etc. will be present when the image of the source alone is produced and thus will not appear in the final image after digital subtraction.

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Furthermore, by varying the energy of the x-ray field with multiple exposures, back scatter tomography is made possible, wherein a 3-D image of the object can be constructed.

5   The measured inherent sensitivity of cells of a detector vary by about 20%. Furthermore, it has been discovered that the sensitivity of the detector 14 is varied by prolonged exposure to x-radiation. That is, as cells of the detector are exposed, they become more
10 sensitized to radiation in the future. Thus, detectors which receive less radiation dose because they are shadow shielded by an object are less sensitized than those exposed to direct or less attenuated radiation. Therefore, the relative sensitivity of each detector
15 varies according to its own history of exposures. X-ray applications with typically useful image gray scales do not allow such large sensitivity variations and thus require sensitivity normalization. Such normalization can be carried out using the following
20 technique. For the following discussion it is assumed that the detector comprises one half of a basic IS32 detector configuration, i.e. a 128 x 256 cell array contained in a spatial area 876.8 x 4420 microns. The cells are arranged in 128 rows and 256 columns.
25 Presuming that the cells are illuminated by a source which generates a uniform intensity field at the sensor, a distribution of "exposure times", t, will be obtained to register a cell discharge. The individual cells are denoted by $t(m,n)$ where m is $1,2,3,...,M$ and
30 n is $1,2,3,...,N$. The letter m signifies the row index and the letter n signifies the column index. In the present example, M is 128 and N is 256. Denoting the average value of this array of 32,768 values by t^*, note that the 20% variability discussed above means
35 that the difference of $t(m,n)$ and t^* to be as large as $0.2t^*$. The cell array normalization factors can be

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defined by f(m,n) equals the ratio of t* to t(m,n).
Thus we can generate the array f(m,n) where m is
1,2,...,M and n is 1,2,...,N. In measuring the set of
values t(m,n) we must set a time increment used to step
5   through the cells. Certainly this increment should be
less than 0.2t* or the measurement will be of little
value.

    Where gray scale is produced by sequential
exposures of the object with different exposure times,
10   the gray scale will be specified by a set of contiguous
exposure time intervals. Denote the increasing bounds
of these intervals by t(j) where j is 0,1,2,...,J.
This defines a J-gray scale image mode (typically J is
8, 16, 32 or 64) bounded by a low exposure time of t(0)
15   and a high exposure time of t(J). The j-th shade of
gray is bounded by exposure times of t(j-1) and t(j).
The image intensity measured by the radiation field at
sensor locations varies inversely with j.

    In a particular image obtained with the J-gray
20   scale just described, an array of values t(m,n) will be
obtained where m is 1,2,...,M* and n is 1,2,...,N*.
The values of M* and N* are respectively equal to or
less than M and N depending on either the fraction of
cells used, or cell spatial averaging procedures. This
25   array of cell exposure times contains the image
information. Since image exposure times are only at
the specified values t(0), t(1),...,t(J) the image
values of t(m,n) will fall in one of the discrete J
30   gray scale intervals and no further detailed knowledge
can be presumed. Each cell response to the image must
be corrected by the sensitivity normalization f(m,n).
There are a variety of ways that this correction can be
applied.

35   One method of carrying out this correction is
shown by the program of Figure 8. This program is run
to read each cell at each exposure. In other words,

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the first exposure time t(0) may be 1.1 sec, the second
exposure time t(1) may be 1.2 sec, etc. After each
exposure, all of the cells t(m,n) are read and the
information output from each cell is stored in a
5   matrix.

In Figure 8, the program is started at step 120
and the variables are initialized at step 122. This
includes setting m and n equal to zero. At step 124,
the value of the first cell t(m,n) is read and the
10   normalization factor for that cell is looked up at step
126. At step 128, the gray scale interval j is
determined. For example, if the cells are being read
after the first exposure and the exposure time is 1.1
sec, the gray scale interval j would be 1.1-1.2 sec. If
15   the second exposure time is 1.1 sec, the second gray
scale interval would be 1.1-1.2 sec, etc.

At step 130, an exposure time t*(j)
characteristic of the j-th interval for each j is
identified. For example, t*(j) could be the simple
20   average; one-half of the sum of t(j) and t(j-1). In
other words, if the interval 1.1-1.2 sec is the current
interval j, it is necessary to assign a single value
t*(j) to the exposure time. This value may be, e.g.,
1.15 sec. Then, at step 132, for each image spatial
25   location (m,n) which has an image value t(m,n) in gray
scale interval j, calculate f(m,n)t*(j) which yields
the corrected value t*(m,n). This value: 1) falls in
interval j and then the assignment of gray shade j for
(m,n) is maintained; 2) falls in the same other
30   interval j' and then the assignment of corrected gray
shade j' for (m,n) is accomplished; 3) is less than
t(0) and then (m,n) is assigned "white"; or 4) is
greater than t(J) and then (m,n) is assigned "black".
The determination of the gray scale interval for
35   t*(m,n) is made at step 134 and the actual assignment
of the gray scale is made at step 136. Clearly the

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choice of $t^*(j)$ and the validity of the procedure is
less uncertain when the intervals $t(j-1)$ to $t(j)$ are
"small". Small is relative to expected cell
sensitivity variation over the image in question. At

5   step 138, a determination is made as to whether all
cells have been read. If the answer is "no" the values
of m and n are incremented at step 140 according to a
predetermined schedule and the value of the next cell
is read. After all the cells are read, the program

10 ends at step 142.

The foregoing description is provided for
purposes of illustrating the present invention but is
not deemed to be limitative thereof. Clearly, numerous
additions, substitutions and other changes can be made

15 to the invention without departing from the scope
thereof as set forth in the appended claims.

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The embodiments of the invention in which an exclusive property or privilege is claimed are defined as follows:

1.   An x-ray imaging system comprising:

an x-ray source for producing an x-ray field;

an x-ray detector comprising a solid state integrated circuit having a silicon substrate and having a plurality of charge storage devices, and circuit means for placing a charge on said charge storage devices, said charge storage devices being disposed in an x-ray permeable material, and said detector being positioned in said x-ray field such that said charge is dissipated by secondary radiation produced by interaction between said x-ray field and said silicon substrate of said solid state integrated circuit.

2.   An x-ray imaging system as claimed in claim 1 wherein said detector is placed between said source and an object to be viewed such that said x-ray field passes through said detector, impinges on said object and forms an image of said object in said detector by back scatter from said object.

3.   An x-ray system as claimed in claim 2 including means for subtracting an image of said source from the image produced in said detector.

4.   An x-ray imaging system as claimed in claim 1 wherein said charge storage devices are divided into groups to form pixels, each pixel having a plurality of charge storage devices and means for varying the sensitivity of said charge storage devices in a single pixel to provide a gray scale.

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5. An x-ray imaging system as claimed in claim 1 including a support housing for said detector, said support housing having means for removably receiving said detector.

6. An x-ray imaging system as claimed in claim 5 wherein said detector has a thickness on the order of 1/2 mm.

7. An x-ray imaging system as claimed in claim 1 wherein each of said charge storage devices is contained within a cell having a greatest dimension of approximately 10 microns.

8. An x-ray imaging system as claimed in claim 1 wherein said integrated circuit comprises a dynamic random access memory.

9. An x-ray imaging system as claimed in claim 1 wherein said circuit means comprises transistors for connecting said charge storage devices to a voltage source.

10. An x-ray imaging system as claimed in claim 1 including means for moving one of said x-ray source and said detector relative to the other by a distance approximating the distance between charge storage devices in said detector.

11. An x-ray imaging system as claimed in claim 1 including means for normalizing the soak times required for discharging all of the charge storage devices which are intended to discharge simultaneously.

12.   An x-ray imaging system as claimed in claim 11
wherein said normalization means comprises means for
storing a different normalizing factor for each of said
cells.

13.   An x-ray imaging system as claimed in claim 1
including a plurality of x-ray detectors positioned
over one another such that the cells of each x-ray
detector are staggered with the cells of the other
x-ray detectors such that any one cell of any one x-ray
detector is positioned between two cells of another
x-ray detector.

14.   An x-ray imaging system as claimed in claim 1
including means for exposing the detector to said
source for sequential, varying lengths of time to
produce a plurality of images, and adding said images
together to provide a gray scale.

15.   An x-ray imaging system as claimed in claim 14
wherein the intervals between said sequential varying
length exposure times are assigned different gray scale
values.

16.   An x-ray imaging system as claimed in claim 4
wherein said varying means comprises different
thickness oxide layers in said charge storage devices.

17.   An x-ray imaging system as claimed in claim 4
wherein said varying means comprises different
comparison voltage levels for determining the charge on
said charge storage devices.

18.   An x-ray imaging system as claimed in claim 1
wherein each said charge storage device forms a single
pixel of an image formed on said detector.

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19.   A method of detecting an x-ray image comprising:

storing charges on a plurality of charge storage devices formed in an integrated circuit material, said charge storage devices being formed by a junction between a semiconductor substrate and an oxide layer, and a metal surface formed on said oxide layer; exposing said integrated circuit to an x-ray field such that said x-ray field interacts with said silicon substrate to produce free electrons to reduce the charge on said charge storage devices.

20.   A method according to claim 19 including producing an image having a gray scale by sequentially exposing said integrated circuit with different exposure times and digitally adding the images obtained at each of said exposure times.

21.   A method according to claim 19 comprising producing a back scattered image by positioning said integrated circuit between said source and an object to be viewed, causing said x-ray field to pass through said integrated circuit and form an image in said integrated circuit through back scattering from said object.

22.   A method according to claim 19 including normalizing the outputs from said charge storage devices to compensate for different sensitivites of said charge storage devices.